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Compensatory strategies during manual wheelchair propulsion in response to weakness in individual muscle groups: A simulation study

Jonathan S. Slowik [a], Jill L. McNitt-Gray [b,c], Philip S. Requejo [d,e], Sara J. Mulroy [d], Richard R. Neptune [a,*]

[a] Department of Mechanical Engineering, The University of Texas at Austin, Austin, TX, USA
[b] Department of Biomedical Engineering, The University of Southern California, Los Angeles, CA, USA
[c] Department of Biological Sciences, The University of Southern California, Los Angeles, CA, USA
[d] Pathokinesiology Laboratory, Rancho Los Amigos National Rehabilitation Center, Downey, CA, USA
[e] Rehabilitation Engineering, Rancho Los Amigos National Rehabilitation Center, Downey, CA, USA

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ABSTRACT

Background: The considerable physical demand placed on the upper extremity during manual wheelchair propulsion is distributed among individual muscles. The strategy used to distribute the workload is likely influenced by the relative force-generating capacities of individual muscles, and some strategies may be associated with a higher injury risk than others. The objective of this study was to use forward dynamics simulations of manual wheelchair propulsion to identify compensatory strategies that can be used to overcome weakness in individual muscle groups and identify specific strategies that may increase injury risk. Identifying these strategies can provide rationale for the design of targeted rehabilitation programs aimed at preventing the development of pain and injury in manual wheelchair users.
Methods: Muscle-actuated forward dynamics simulations of manual wheelchair propulsion were analyzed to identify compensatory strategies in response to individual muscle group weakness using individual muscle mechanical power and stress as measures of upper extremity demand.
Findings: The simulation analyses found the upper extremity to be robust to weakness in any single muscle group as the remaining groups were able to compensate and restore normal propulsion mechanics. The rotator cuff muscles experienced relatively high muscle stress levels and exhibited compensatory relationships with the deltoid muscles.
Interpretation: These results underline the importance of strengthening the rotator cuff muscles and supporting muscles whose contributions do not increase the potential for impingement (i.e., the thoracohumeral depressors) and minimize the risk of upper extremity injury in manual wheelchair users.

1. Introduction

Over half of all manual wheelchair users will develop upper extremity pain and injury at some point in their lifetime (e.g., Finley and Rodgers 2004), which can be highly debilitating and lead to a decrease in independence and quality of life (e.g., Gutierrez et al. 2007). This high incidence of pain and injury is correlated with the considerable physical demand placed on the upper extremity during wheelchair propulsion (e.g., Curtis et al. 1999), as significant intermuscular coordination is needed to generate the mechanical power necessary to propel the wheelchair while maintaining joint stability (e.g., Rankin et al. 2010, 2011, 2012; van der Helm and Veeger 1996).

Many different combinations of muscle forces can produce the same net joint moments and generate the required mechanical power (e.g., Pandy and Andriacchi 2010). Although there is some uncertainty in how the neuromuscular system selects a particular combination of muscle forces to perform a given movement task, most theories suggest that the relative levels of force-generating capacity in individual muscles influence the selection (Erdemir et al. 2007). Muscle weakness (or decrease in the capacity to generate force) can be influenced by a number of factors including fatigue and neurological deficits (Requejo et al. 2008).

Muscle fatigue can result from a number of mechanisms, but it is generally quantified as a transient reduction in the force capacity of a muscle due to sustained physical activity (Enoka and Duchateau 2008). In order to fulfill specific task requirements, fatigue may occur at different rates in individual muscles and resulting fatigue-related changes in musculoskeletal loading may lead to injury (e.g., Kumar 2001). However, the overall effect of fatigue on wheelchair propulsion biomechanics is not well understood, as one study concluded that fatigue may lead to potentially harmful changes in propulsion mechanics (Rodgers et al. 1994), while others have suggested that during an

* Corresponding author at: Department of Mechanical Engineering, The University of Texas at Austin, 204 E. Dean Keeton Street, Stop C2200, Austin, TX 78712, USA.
E-mail address: rneptune@mail.utexas.edu (R.R. Neptune).

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extended period of propulsion, individuals may actually make beneficial adjustments to their propulsion mechanics to mitigate the increased risk of injury (Rice et al. 2009). A number of inverse dynamics-based analyses have found that during manual wheelchair propulsion, the highest net joint moments and powers are generated at the shoulder, suggesting that the glenohumeral joint may be the most at risk for overuse injury (e.g., Rodgers et al. 1994; Veeger et al. 1991). These analyses also identified small fatigue-related shifts in joint power from the glenohumeral joint to more distal joints (Rodgers et al. 2003). Recently, Qi et al. (2012) found that electromyography intensity increases with fatigue and suggested that fatigue may contribute to imbalances between the contact and recovery phase muscles. However, the effect of fatigue in individual muscles on propulsion mechanics has remained relatively unexplored.

Muscle weakness can also result from neurological deficiencies due to injury or disease and ensuing neuromuscular changes, such as denervation and atrophy (e.g., Thomas and Zijdewind 2006). Furthermore, the breadth and magnitude of these reductions can vary based on the specific impairment or injury level. For example, a person with paraplegia will likely be able to produce larger forces with their triceps and pectoralis major muscles than a person with tetraplegia (e.g., van Drongelen et al. 2006). However, despite these differences, glenohumeral joint kinematic patterns and net joint moments during wheelchair propulsion have been shown to be remarkably similar across different spinal cord injury levels (Kulig et al. 2001; Newsam et al. 1999).

Although different combinations of muscle forces can produce the same net joint moments and minimize the effect of individual muscle weakness on propulsion mechanics, it is important to understand the potential compensatory strategies used by the neuromuscular system, as the resulting combinations of muscle forces may put the upper extremity at a higher risk for the development of pain and injury. The potential for injury has been highlighted in previous studies showing that joint instability due to larger forces from the deltoid relative to the humeral head depressors (i.e., rotators and adductors) may lead to subacromial impingement (e.g., Burnham et al. 1993; Sharkey and Marder 1995) and that other unbalanced combinations of forces can lead to dislocation (Labriola et al. 2005). These conditions are also related to the high prevalence of rotator cuff tears among manual wheelchair users (Morrow et al. 2014).

Forward dynamics simulations have been shown to be an effective tool to advance our understanding of intermuscular coordination during wheelchair propulsion (e.g., Rankin et al. 2011; Zajac et al. 2002). Potential compensatory strategies in response to individual muscle weakness can be revealed through analyzing the resulting shifts in individual muscle activation or power generation. A similar approach has previously been used to determine the effect of muscle weakness during steady-state walking (Goldberg and Neptune 2007; Jonkers et al. 2003; van der Krogt et al. 2012). Forward dynamics simulations can also be used to examine specific measures of upper extremity demand, such as muscle stress, to help identify muscles that may be placed at risk for overuse injuries (Rankin et al. 2012).

The purpose of this study was to use musculoskeletal modeling and forward dynamics simulations of wheelchair propulsion to identify potential compensatory strategies necessary to overcome weakness in individual muscle groups and highlight those strategies that could lead to the development of upper extremity pain and injury. The results of this study can provide rationale for the design of targeted rehabilitation programs aimed at minimizing the development of pain and injury in manual wheelchair users.

2. Methods

2.1. Musculoskeletal model

The upper extremity musculoskeletal model and dynamic optimization framework used in this study to generate the simulations of manual wheelchair propulsion have been previously described in detail (Rankin et al. 2010, 2011). The musculoskeletal model was developed using SIMM (Musculographics, Inc., Santa Rosa, CA, USA) based on the work of Holzbaur et al. (2005) and consisted of segments representing the trunk and right upper arm, forearm and hand. There were six rotational degrees-of-freedom representing trunk lean, shoulder plane of elevation, shoulder elevation angle, shoulder internal–external rotation, elbow flexion–extension, and forearm pronation–supination. The shoulder angles were the thoracohumeral angles, while the scapulohumeral rhythm was defined using regression equations based on cadaver data (de Groot and Brand 2001). Full-cycle trunk lean and contact-phase hand translations were prescribed based on experimentally measured kinematic data. The dynamic equations of motion were generated using SD/FAST (Parametric Technology Corp., Needham, MA, USA). Twenty-six Hill-type musculotendon actuators, governed by intrinsic muscle force-length-velocity and tendon force-strain relationships, represented the major upper extremity muscles crossing the shoulder and elbow joints (e.g., Slowik and Neptune 2013). Each actuator received a distinct excitation signal except the two sternocostal pectoralis major actuators, the three latissimus dorsi actuators, and the two actuators representing the lateral triceps and anconeus. Within each of these groups, the actuators received the same excitation signal, resulting in a total of 22 excitation groups. Muscle excitation–activation dynamics were modeled using a first order differential equation (Raasch et al. 1997) with muscle-specific activation and deactivation time constants (Happee and van der Helm 1995; Winters and Stark 1988). The musculotendon lengths and moment arms were determined using polynomial regression equations (Rankin and Neptune 2012) and the product of the appropriate muscle moment arm and force determined the muscle moment that was applied to each joint. In addition, passive torques were applied at the joints to represent ligaments and other passive joint structures that limit extreme joint positions (Davy and Audu 1987).

2.2. Simulation and optimization framework

Each muscle excitation pattern was generated using a bimodal pattern defined by six parameters (Hall et al. 2011), resulting in a total of 132 optimization parameters. A simulated annealing optimization algorithm (Goffe et al. 1994) was used to identify the excitation parameters that produced a simulation that best emulated the group-averaged experimental propulsion data (i.e., joint angle and 3D handrim force profiles; see Section 2.3 below) using an optimal tracking cost function (Neptune et al. 2001). An additional term was included in the cost function that minimized the square of muscle stress to prevent unnecessary co-contraction.

Based on a combination of anatomical location and muscle function, the musculotendon actuators were assigned to 12 muscle groups for analysis (Table 1). An initial simulation was generated using a set of baseline isometric muscle force values in the musculoskeletal model based on anatomical studies (Holzbaur et al. 2005; Table 1). These values were then systematically reduced by 50% one group at a time with the remaining groups left unaltered. The 50% reduction was deemed large enough to provide a meaningful difference and realistic in that it falls within the range of values reported in the literature due to fatigue from similar submaximal tasks (e.g., Enoka and Duchateau 2008). The excitation pattern of the weakened group was constrained to remain at the baseline values so that it could not compensate for itself, and the muscle excitation patterns of the remaining groups were re-optimized in order to restore the propulsion mechanics that emulated the experimental propulsion data, resulting in an additional 12 simulations.

2.3. Experimental data

To provide tracking data for the dynamic optimization, experimental data from twelve experienced male manual wheelchair users with